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# Gait and electromyographic analysis of patients recovering after limb-saving surgery

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## Abstract

**Objective.** Control of gait after limb-saving surgery.

**Design.** Case series study.

**Background.** At the moment little is known about adaptations in patients' gait after limb-saving surgery.

**Methods.** Nineteen patients who underwent limb-saving surgery at least 1 yr earlier and 10 normal subjects were studied during treadmill walking. The main outcome measures were walking speed, step parameters and angular displacement of both legs and EMG of the biceps femoris, rectus femoris and medial gastrocnemius in the affected leg.

**Results.** Preferred walking speed in the patients was lower than in the controls (0.7 versus 1.1 m/s). Furthermore, stance phase of the non-affected leg was lengthened. All patients showed reduced stance phase knee flexion in the affected leg, while during the swing phase no difference was seen. The EMG signals of the rectus femoris and biceps femoris show changes, which are related to the location of surgery.

**Conclusions.** The results showed that the gait pattern of the patients differed compared to normal gait. The reduced stance phase knee flexion in the hip group is based on a high degree of co-contraction between quadriceps and hamstring activity, while in the knee group this is based on the quadriceps avoidance pattern. The finding that there is still side-to-side asymmetry indicates that there is no complete reorganisation following the massive loss of input and output of the leg. It is possible that some reprogramming of the locomotor process occur.

## Relevance

Gait and electromyographic analysis are essential for the quantitative assessments of the functional outcome in this type of surgery. © 2000 Elsevier Science Ltd. All rights reserved.

**Keywords:** Gait; Gait adaptations; EMG; Limb-saving surgery; Oncology; Hip; Knee

## 1. Introduction

Until now almost nothing is known about the recovery of functional gait after limb-saving surgery (LSS), which is a technique employed in patients with malignant bone or soft tissue tumours in the extremities [1]. Most of the studies have been focussed on disease- or impairment-specific variables or have presented only very global information concerning the asymmetry between the affected and non-affected leg during gait,

however only a few studies are focussed on the functionality of gait during daily activities [2–6].

Reduced stance phase knee flexion is a common finding after total knee replacement [7–9]. Until now it has only been reported in LSS patients after distal femoral knee prosthesis implants [4]. It is logical to relate the changes after distal femoral knee prosthesis to local adaptations due to the massive resection of the joint and the ligaments in these patients. However, we also see this pattern in patients after LSS for the hip. In the recent literature, there are no studies addressing this item, not even in patients with total hip replacement. The crucial question is how specific the reduced stance phase knee flexion pattern is and what is the control

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mechanism by the motor system. It is important to note that after LSS the limb has been changed structurally. In these patients, massive bone components are removed and replaced by a prosthesis. Little is known of how the motor system compensates for such a totally changed architecture of the leg. One can speculate about a potential reprogramming of the locomotor process. This reprogramming may originate from higher neuronal centres. These compensations might be relearned responses. The primary aim of the present study is to reveal information about the compensations by quantifying the changes in temporal gait parameters, muscle activity and knee kinematics during treadmill gait in patients with LSS for the knee or the hip. It is argued here that the employed measurement procedure will give insight into the strategy these patients use in order to overcome the motor deficits due to the surgery.

## 2. Methods

### 2.1. Subjects

Nineteen patients, 12 male and 7 female, with a mean age of 45 yr (range 21–80), who were all treated for malignant bone tumours in the lower extremity participated in the study (Table 1). The study was approved by the local committee for medical ethics and all subjects gave their informed consent. Two groups were defined: a group that underwent knee surgery ( $n = 9$ ) and a group that underwent hip surgery ( $n = 10$ ). The knee surgery included a reconstruction with a distal femoral knee prosthesis and a partial resection of the quadriceps muscles, like medial or lateral vastus. The hip surgery means a reconstruction with proximal femoral or saddle prosthesis and a partial resection of the gluteal muscles or tensor fascia lata. None of the patients suffered from neurological disease, pathology like back pain or arthritis in the contra-lateral leg, which could influence their performance. All patients had an active range of motion, which should allow normal gait. 10 healthy controls, four male and six female, were included with a mean age of 37 yr (range 22–61).

### 2.2. Procedure

The patients were measured in a gait laboratory, 12–24 months after surgery. The following gait parameters were determined: preferred walking speed, stride time, stance time, swing time, double-limb support. Thin insole footswitches were placed in the shoes to detect heel-strike and toe-off of both feet. The knee angles of both the knees were measured using laterally placed electrogoniometers (Penny and Giles type m180). To filter the data of the electrogoniometers, a second-order Butterworth filter with a cutoff frequency of 10 Hz was used.

Table 1  
Gait parameters

|          | N  | Age (date of measurement) |         | Walking speed (m/s) |        | Stride time (s) |        | Stance non-aff (% of step cycle) |       | Stance aff (% of step cycle) |       | Double support time (% of step cycle) |      |
|----------|----|---------------------------|---------|---------------------|--------|-----------------|--------|----------------------------------|-------|------------------------------|-------|---------------------------------------|------|
|          |    | Mean                      | (range) | Mean                | (SD)   | Mean            | (SD)   | Mean                             | (SD)  | Mean                         | (SD)  | Mean                                  | (SD) |
| Hip      | 10 | 47                        | (21–66) | 0.6                 | (0.3)  | 1.62            | (0.89) | 62                               | (6.6) | 55                           | (5.1) | 13                                    | (5)  |
| Knee     | 9  | 45                        | (37–80) | 0.8                 | (0.3)  | 1.41            | (0.28) | 63                               | (2.6) | 59                           | (1.8) | 14                                    | (2)  |
| Total    | 19 | 45                        | (21–80) | 0.7                 | (0.3)  | 1.50            | (0.60) | 62                               | (4.8) | 57                           | (4.2) | 14                                    | (3)  |
| Patients |    |                           |         |                     |        |                 |        |                                  |       |                              |       |                                       |      |
| Controls | 10 | 37                        | (22–61) | 1.1                 | (0.08) | 1.10            | (0.06) | 57                               | (2.6) |                              |       | 11                                    | (1)  |

Bipolar surface electromyographic (EMG) activity of the affected leg was recorded by means of surface electrodes placed on the belly of the following muscles: biceps femoris (BF), rectus femoris (RF) and the medial gastrocnemius muscle (MG). The skin was prepared by shaving and mildly abrading the electrode site. Inter-electrode distance was less than 2 cm and no test was initiated if the measured skin impedance exceeded 5 k $\Omega$ . The EMG signal was (pre-) amplified (by a factor in the order of  $10^4$  to maximally  $10^6$ ), high-pass filtered ( $>3$  Hz), full-wave rectified, low-pass filtered ( $<3000$  Hz) and transferred on-line using an AD-converter (sampled at 500 Hz). To enable a proper inter-subject comparison of the amplitudes, the data of each muscle were normalised with respect to the maximal EMG activity during the step cycle.

Before the experiment started, all the subjects were allowed to walk on the treadmill to get used to the peculiarities of treadmill walking and to find their preferred walking speed. The measurements took place within a time interval of 100 s. Each measurement consists of at least 80 strides (heel-strike to heel-strike).

To avoid masking of important features of individual responses, all individual trials and subjects-data were examined prior to averaging. The parameters were averaged over the complete strides. Step parameters were calculated and averaged from the footswitch data. From the kinematics data of the knee angles, we calculated the peak range of motion during the stance and swing phase. The angular velocity (AV) was the first-order differential from the knee angle (angular velocity =  $\Delta$  (angular displacement)/ $\Delta t$ ; [deg/s]). The analysis was limited to the regions of the maximum and minimum AV in the swing phase. The EMG activity was calculated for each muscle; the data were normalised, averaged and expressed as a percentage of step cycle. Statistical analysis of time-dependent variables (e.g. knee displacement and AV of the knee) was limited to regions of relative maximum or minimum velocity, because these were the regions in which differences between subject groups were most apparent. A correlation coefficient was determined to measure the relationship between the variables. A two-way analysis of variance (ANOVA) was performed to analyse the difference between controls and patients and between the two patients groups. An  $\alpha$ -level of significance of 0.05 was selected for all statistical tests.

### 3. Results

The gait parameters are depicted in Table 1. The mean preferred walking speed in the patients was 0.7 m/s (SD 0.3), while it was 1.1 m/s (SD 0.08) for the controls. The mean stride duration in the patient group was 1.5 s (SD 0.6), which was longer, compared to the healthy

subjects (mean 1.1 s, SD 0.06) (ANOVA,  $P < 0.05$ ). All the patients showed a shortening of the stance phase of the affected leg compared to the non-affected leg (respectively, 57 (SD 4.2%) of the cycle as compared to 62 (SD 4.8%); ANOVA,  $P < 0.05$ ). The stance phase of the non-affected leg in patients was longer (62, SD 4.8% step cycle) as compared to the controls (57, SD 2.6% step cycle; ANOVA,  $P < 0.05$ ). For the double-limb support time, we found slightly elevated values for the patient groups (mean 14%, SD 3), but no significant difference with the controls (mean 11%, SD 1). For none of the parameters mentioned was there a difference between the hip and knee group.

#### 3.1. Knee angles

The above mentioned gait asymmetry is also reflected in the kinematic parameters of the knee. Fig. 1 shows typical results of averaged EMG activity and angular displacement of the knee in two patients and a healthy control.

One patient had a tumour in the knee and a resection and reconstruction with distal femoral knee prosthesis. The other had a tumour in the hip and a resection and reconstruction with proximal femoral hip prosthesis. In both the knee and the hip patients, the angular displacement of the knee showed less knee flexion during the stance phase. The maximum BF activity in the control subject occurred before initial contact. In the knee patient, this activity shifted towards the beginning of the stance phase. The knee patient showed a prominent extra burst appearing at the end stance. In the hip patient, there was also a prominent burst at the stance–swing transition, but the peak occurred later after the onset of swing. In the control, the RF was mainly active at the initial contact. In the knee patient, the pre- and early swing RF activity (“swing burst”) was dominant, while the end swing and early stance activity (“stance burst”) was much more reduced as compared to the control. In the hip patient, the RF activity started early in the swing phase, reaching its maximum activity in the late swing and initial contact. In the control, the MG activity started during the loading response directly after initial contact and remained active in the mid and terminal stance. Furthermore, in the knee patient, there was exceptionally little MG activity during early stance. The hip patient showed a more normally MG burst pattern, but still with a shorter burst duration than the control.

The mean peak range of motion (RoM) during the stance and swing of the entire patient and the control group are depicted in Fig. 2. In most of the patients, in the knee group (six out of nine), slight hyperextension is seen during the stance. The other three showed less flexion in the stance phase than the controls.

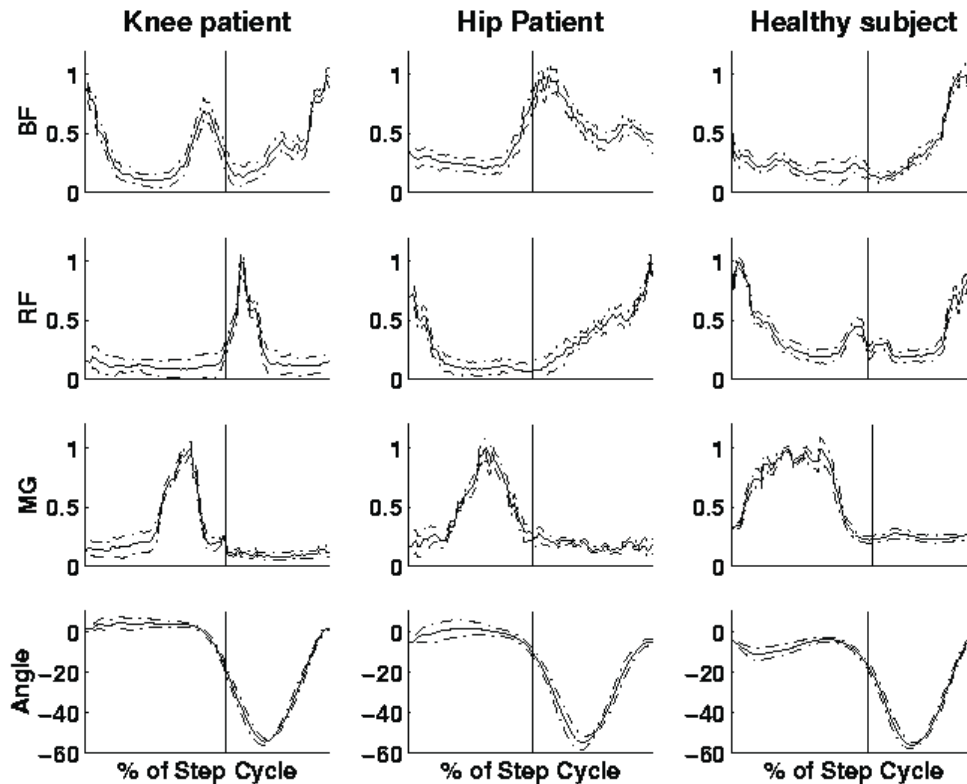


Fig. 1. The mean and SD of the BF, RF MG and knee angle in the affected leg of two individual patients and one healthy control.

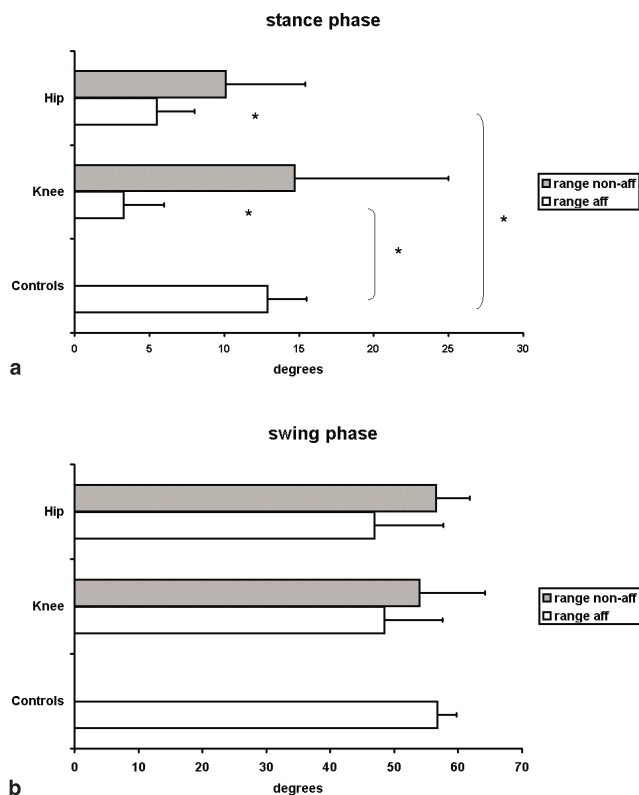


Fig. 2. The peak range of motion of the knee of patients and healthy controls, during stance phase (a) and swing phase (b). Significant difference is marked with \*.

In the patients who underwent knee surgery, the mean RoM of the knee during the stance phase was  $15^\circ$  (SD 5) in the non-affected leg and  $3^\circ$  (SD 2.7) in the affected leg. Patients after hip surgery showed respectively  $10$  (SD 4.2) and  $6^\circ$  (SD 2.5). In the healthy subjects, we recorded  $13^\circ$  (SD 2.6). In all the patients, the RoM in the stance phase was significantly lower in the affected leg compared to the non-affected leg (ANOVA,  $P < 0.05$ ) and compared to the controls (ANOVA,  $P < 0.05$ ). The correlation between the walking velocity and the RoM during the stance phase in the affected and non-affected leg was weak and not significant (respectively,  $r = 0.33$   $P = 0.12$ ;  $r = 0.06$   $P = 0.09$ ). The RoM during the swing phase in general did not show a difference between the knee and hip patients (ANOVA,  $P > 0.05$ ) and between the patients and controls (ANOVA,  $P > 0.05$ ). The correlation with the walking velocity and the RoM during swing was high for the affected leg ( $r = 0.7$   $P = 0.05$ ) and weak for the non-affected leg ( $r = 0.46$   $P = 0.075$ ).

To have a closer look at the difference in the swing phase, the AV of the knee of the same three individuals as shown in Fig. 2 is depicted in Fig. 3. In general, a lower AV was seen in both the legs of the patients as compared to the controls.

In Fig. 4, the peak AV during the swing phase for the total patient and control group is depicted. Compared to the controls, the results in flexion and extension for

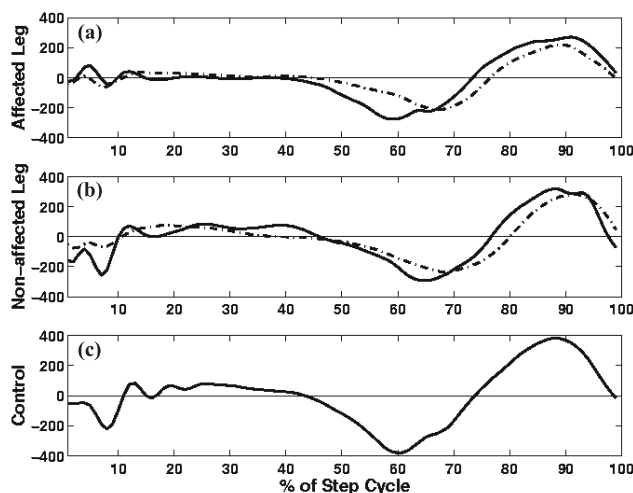


Fig. 3. The angular velocity of the knee in two patients. The patient with the affected knee is depicted with the solid line (—), the patient with the affected hip is depicted with the dotted line (---), affected leg (a), non-affected leg (b) and a control, depicted with a solid line (c).

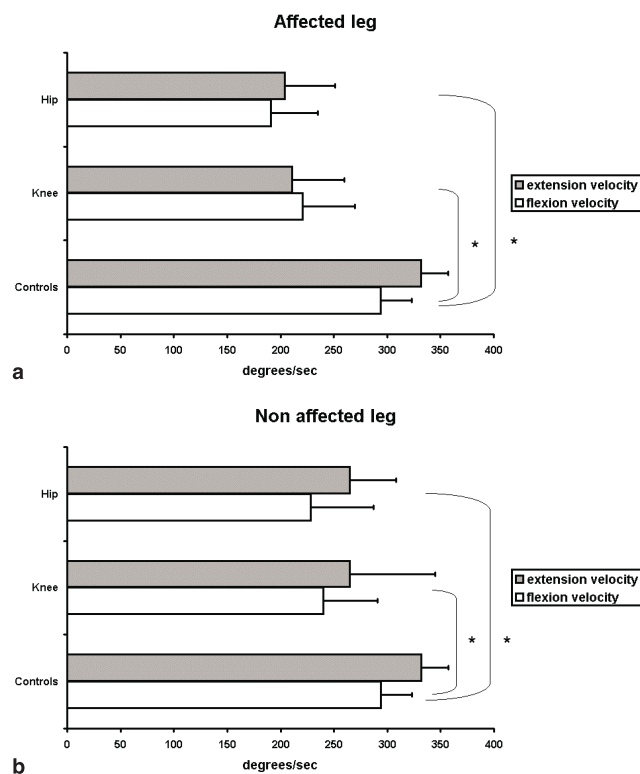


Fig. 4. The peak angular velocity in flexion and extension during swing of the patients affected leg (a) and non-affected leg (b). Significant difference is marked with \*.

both the legs were significantly lower (ANOVA,  $P < 0.05$ ). For the knee group, the AV for the affected and non-affected leg was 221 (SD 49) and 240 deg/s (SD 51), respectively. For the hip group, it was 191 (SD 44) and 228 deg/s (SD 59), respectively. Although the flexion velocity in the non-affected leg is slightly higher for both

the groups, we did not find a significant difference between the affected and non-affected leg (ANOVA,  $P > 0.05$ ). Note that there was also no significant difference between limb velocities of the patient groups (ANOVA,  $P > 0.05$ ). The peak extension velocity showed the same tendencies. For the knee group, the AV for the affected and non-affected was 211 (SD 49) and 265 deg/s (SD 80). For the hip group, this was respectively 240 (SD 47) and 265 deg/s (SD 43). One consistent finding was that in the patients with knee surgery, on the affected side, the flexion AV was higher than the extension AV (student- $t$ ,  $P < 0.05$ ). This finding differs from the hip group and the healthy controls in which the peak AV is slightly higher in extension.

The correlation between the AV and the walking speed was high and significant for the extension velocity in swing phase of the non-affected leg. In the affected leg, there was only a modest correlation (respectively,  $r = 0.73$   $P < 0.05$ ;  $r = 0.65$   $P < 0.05$ ). For the AV of the flexion in the swing phase, the correlation was weak in both the affected and non-affected legs, respectively ( $r = 0.65$   $P < 0.05$ ;  $r = 0.55$   $P < 0.05$ ).

### 3.2. Changes in EMG patterns of the affected leg

The results of the mean EMG activity of the main muscles that control the knee have been summarised in Fig. 5 for both patient groups. In these means, the difference between the patients tended to be smoothed compared with the individual examples in Fig. 1, yet the same consistent differences were seen.

For example, the BF in the knee group showed an activity shift of the stance–swing burst towards the swing phase as compared to control subjects, who showed this burst always before the start of the swing phase (Fig. 5(a)). In the BF, two main changes were seen. First, both in the knee and the hip group, the two main bursts of activity occurred at the phase transitions (swing–stance and stance–swing), while they preceded the transitions in the controls. Second, the two bursts (end stance, end swing) were more similar in amplitude in the both knee and hip patients, while in the controls the end swing burst was considerably larger than the end stance burst (Fig. 5(d)). Similarly, in the RF the normal large burst at end of swing was much reduced in the knee and hip patients as compared to the controls (Fig. 5(b) and (h)). In the hip patient group, the RF showed more activity during stance and swing (Fig. 5(e)). For the MG, the peak activity was shifted towards end stance in both patient groups (Fig. 5(c), (f) and (i)).

## 4. Discussion

The aim of the present study was to investigate the changes in knee kinematics and muscle activity during

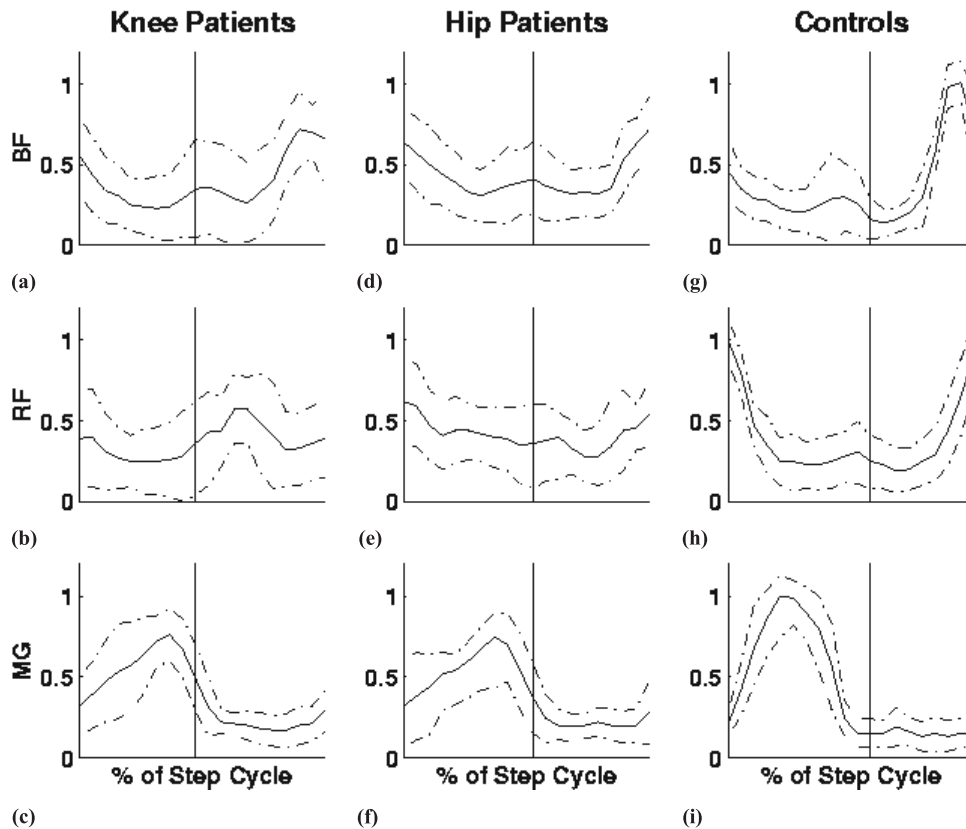


Fig. 5. BF, RF and MG activity in the affected leg of knee and hip patients and the healthy subjects. The EMG is normalised for each individual. The mean EMG of the three groups is depicted by the straight lines, the dotted lines indicate the SD and the EMG to percent of step cycle.

treadmill gait in LSS patients. Although the patient group is rather heterogeneous, the results showed that the gait pattern of the patients differed consistently from normal gait patterns. The patients walked with a lower preferred speed and a lengthened stride time, while the double-limb support time was normal. These findings were similar to other studies, not only after LSS but also after total hip or knee replacement [3–5,7–11]. Our results show that the duration of the stance phase of the non-affected leg has lengthened as compared to the duration of the stance phase of controls, while the stance phase of the non-affected leg was similar to that of the controls. This presumably is consistent with taking over part of the loading function of the affected leg. The non-affected leg had to provide support, which lasted long enough to allow the swing to be made by the fast leg. In agreement with the present study, the data of Dietz et al. showed that in limping, the duration of the swing phase was quite flexible [12].

One striking result was that in the knee angles some small but consistent changes were seen as compared to the controls. In general, from a kinematic point of view, the patients were able to walk with a smooth, unbroken stride and the main changes in comparison with the controls were seen during the stance phase. Here the result of our study was that no differences could be

found between the knee and the hip group in the kinematic parameters of the knee. This is striking, particularly since the damage to the locomotor apparatus was very different and the main changes one might have expected concerned knee angles. Furthermore, the reduced knee flexion in the stance phase cannot be explained by the lower walking speed in the patient group since there was no correlation between the RoM in the stance phase and the walking speed. The extensive resection of the muscles and ligaments at the knee or hip level caused a loss of proprioceptive feedback, which normally is very important for the control of locomotion [13–15]. The loss of proprioceptive input may have contributed to the reliance on 'safe' (hyper)extension in the knee during stance. Previous studies indicated that such a gait might result from joint stabilising co-contraction of the flexor and extensor muscles [4]. The present EMG data are in agreement with this conclusion for the hip group. Here a co-contraction between the hamstrings, quadriceps and gastrocnemius is seen. However, this is not an explanation for the knee group. Here the EMG changes showed different hamstring and less quadriceps activity during the stance phase. This is in agreement with Winter et al. [15], who showed that during pathologic gait there is less muscular activity, in general. Hence, in early stance less knee flexion is

allowed and the acceptance of body weight could be swift towards early stance phase [3]. This position would reduce the demands on the musculature needed to prevent the knee from collapsing and would reduce the bone–bone contact forces, while full knee extension has the advantage of being the most stable weight-bearing position [15].

To obtain smooth foot clearance during normal gait, a minimum RoM is needed during swing [14,15]. Furthermore, during swing, loading is not an issue and a normal RoM of the knee was present in the patients. The way in which this was achieved was different in both the patient groups. In the knee group, it seems that the RF plays an important role in flexion in the hip to gain a smooth toe-off and to compensate for the reduced knee flexion, while the BF has a task in active flexion of the knee. In the hip group, the BF plays an important role in flexing the knee in the swing phase, while the RF is silenced in this period.

Our observations, that the extension velocity was lower than the flexion velocity during the swing phase of the affected leg in the “knee-group”, is in agreement with the findings of Tsuboyama et al. [3]. The RoM and the AV were reduced in the knee not only on the affected but also on the non-affected side. This can be fully explained by the lower preferred walking speed in the patients when we consider the extension velocity. The flexion velocity in the affected leg also showed a positive correlation with the walking velocity. However, the non-affected leg showed a weak correlation with the walking speed. We suggest that this be explained by the lengthening of the stance phase, which will lead to a shortening and speeding up of the swing phase in this leg.

The crucial question is: are these relearned responses which are controlled by higher neuronal centres or are these lower level peripheral compensation mechanisms based on locomotor deficits. Our and other studies found a lengthening of the stance phase of the non-affected leg as compared to the affected leg [2–5]. Some adaptations, such as the asymmetry in stance phase duration, may be due to low level adjustments. The reason for this compensation could be the fact that essential lower limb muscles are removed and that it is impossible for the control system to re-organise its function. It is also known that load feedback can prolong the stance phase even in reduced decerebrate preparations [12]. Other adaptations may require a higher level of control. For example, the finding that the knee angles were relatively similar on the affected and non-affected side (with the exception of the stance phase angular displacements) suggest that geometrical constancy is aimed for and this is likely to be related to high level programming. A recent study showed that during treadmill walking these patients rely more on visual information and cognitive regulation of the walking [6]. This finding will also contribute to the higher level

control. In conclusion, it is possible that reprogramming of the locomotor process occur. This can be accomplished by altering the patterns of muscle contraction as part of an adaptive locomotor program.

This reprogramming hypothesis has some implications for the clinical management of LSS patients. It provides a possible explanation of why some people adapt, whereas others have difficulties with gait. Since we did not measure the EMG of the non-affected leg, it was not possible to test this hypothesis exactly for the entire gait pattern. Some subjects may and others may not have a sufficient sense of the stability occurring at the knee to fine-tune the balance between the quadriceps and hamstring activities using a higher level of movement coordination. The rehabilitation of these patients should focus on the relearning of coordinated muscle activity. Directions for further research in this particular type of patients should consist of a detailed examination of the activation patterns of lower extremity musculature for the intact and affected leg. This would provide additional insight into the unique motor adaptations made by these patients in both affected and intact legs.

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